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(21) *Patentansökningsnummer* / *Patent application number* — *0400145-9*

(86) *Ingivningsdatum* / *Date of filing* — *2004-01-27*

Stockholm, 2005-02-04

För Patent- och registreringsverket
For the Patent- and Registration Office

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AN ARRANGEMENT AND METHOD FOR ASSESSING JOINTS

Technical Field

The present invention generally relates to an arrangement and a measurement method for assessing joints. More specifically, the present invention relates to an arrangement and a method for measuring joint cartilage quality, such as cartilage thickness, cartilage surface roughness and degree of cartilage fibrillation. The arrangement comprises an arthroscopic probe and means for light source driving/control, light detection, signal processing and presentation.

Prior art

Arthritis is a group of common, chronic diseases with great consequences for the individual patient and society. The prevalence of osteoarthritis, for instance, increases after 50 years of age, with women at greater risk than men. Factors that increase the risk of arthritis are sport and work related injuries and overweight (PEYRON, 1986). Arthroscopy has been used for the diagnosis and therapy of orthopedic disorders since the beginning of the twentieth century. Takagi (TAGAKI, 1918, in ALTMAN and KATES, 1983) modified a pediatric cytoscope to fit the problem area of knee joint visualization which became the birth of this dynamic and rewarding field. The arthroscope is primarily a diagnostic device but therapeutic variants allow the removal of adhesions, intra-articular debris and meniscectomy (JACKSON, 1983). The arthroscope gives visual information from the interior of a joint, transmitted via optical fibres to the human eye. Demands have been raised, though, that a more quantitative approach in diagnostic work would improve the quality of therapeutic decisions. Thickness of cartilage, cartilage surface roughness and the degree of cartilage fibrillation are all parameters of great interest in a more quantitative approach to cartilage diagnostics.

Many researchers have earlier suggested ways for cartilage thickness assessment. Several of the methods suggested require a disarticulation of the joint. Usually, these methods measure the undeformed thickness of the cartilage layer. Armstrong and Mow (ARMSTRONG and MOW, 1982) developed an optical method, applied to an isolated cartilage/bone specimen, in which the cartilage/bone interface was easily detected. Needle probe methods (HOCH et al., 1983; MOW et al., 1989; RÄSÄNEN et al., 1990) measure force and displacement of a sharp needle penetrating the cartilage layer from

which cartilage thickness can be calculated. The needle method does not require an isolated specimen, and *in situ* surfaces can be tested. Jurvelin (JURVELIN et al., 1995) has compared microscopy based measurements, the needle probe method and the ultrasound technique. The linear correlation coefficient between microscopy and the needle probe measurements was 0.97 (n = 80) and 0.91 (n = 45) between microscopy and ultrasonic measurements. Strong correlation was also obtained between the needle probe measurements and the ultrasonic measurements. The difference between the three methods is of the order of 0.1 mm or less at a mean sample thickness of 0.86 mm. The authors conclude that the three different techniques for cartilage thickness measurements are highly related.

In situ cartilage thickness has also been measured with high resolution ultrasound (MODEST et al., 1989; RUSHFELDT et al., 1981). Wayne (WAYNE et al., 1998) utilized a radiographic and image analyzing method for thickness studies in articulated joints. Swann and Seedhom (SWANN and SEEDHOM, 1989) described an improved needle technique for thickness measurements. These authors all question the methods which disrupt the cartilage layer because of the thickness changes caused by dehydration or hydrophilic swelling. However, the authors report an accuracy of $\pm$ 0.012 mm with a repeatability of 1.2% for the needle probe.

Optical coherence tomography (OCT) has recently been suggested as a tool to assess articular cartilage structure and thickness (HERRMANN et al., 1999; DREXLER et al., 2000). OCT is based on interferometry between light from a scanning mirror and the cartilage sample. Hermann et al. report resolutions of 5 – 15 μm and differences between OCT and histological measurements of the order of 7 – 9 %. Drexler et al. suggest that polarisation sensitive OCT (PSOCT) can be advantageous for the quantification of collagen structure changes, associated with osteoarthritis.

Magnetic resonance imaging (MRI) has been increasingly used to assess articular cartilage injuries and arthritis. *In vitro* bovine knees have been examined with MRI (MAH et al., 1990). Signal variations have been noticed in degenerated cartilage (LEHNER et al., 1989). MRI has also been used *in vivo* in a canine arthritis model (BRAUNSTEIN et al., 1990). These studies showed hypertrophic articular cartilage repair and other changes associated with osteoarthritis. MRI has also been used increasingly in human studies (for a review see RECHT and RESNICK, 1994). Many of the studies in humans are knee studies with a focus on the identification of focal defects

and cartilage thickness. In these studies the arthroscopy method is often referred to as the gold standard.

Roentgenological techniques have also been used for cartilage studies. Double-contrast techniques (HALL and WYSHAK, 1980) have been frequently used in thickness studies in relation to sex, weight and height.

Reflection spectroscopy is a well-established method for investigation of the structural/molecular composition of a tissue volume. Light from a broad band light source is brought to impinge on the tissue. The light is absorbed and scattered in the tissue volume. The detected spectral distribution of diffusely scattered light carries information about the molecular/structural composition of the tissue passed by the photons.

Cartilage behaves spectroscopically almost like a sheet of white paper whereas the underlying bone has a very different reflection spectrum. The differences can partly be explained by the haemoglobin content of bone. Bone is perfused with blood as opposed to cartilage which is nutritionally supported from the joint liquor.

The source of inspiration to the present work is the clinically expressed demand to perform cartilage thickness measurements during arthroscopic assessments of joints. A spectroscopic method for this purpose would be easily combined with, or integrated into, an arthroscope that would permit simultaneous conventional arthroscopic investigations with quantitative measurements of cartilage thickness. In a longer perspective, it is probably possible to utilize optical fibre measurements for a variety of other important properties such as bone perfusion, cartilage surface topology and degree of fibrillation (NÖTZLI et al., 1989; HANDLEY et al., 1990; DREXLER et al., 2000).

Summary of the invention

The invention is based on the differences in the optical properties of cartilage and subchondral bone. Bone is perfused with blood whereas cartilage mainly consists of collagen fibers and proteoglycan aggregates. The two substances thus show marked differences in absorption spectrum and scattering. The described arthroscope provides joint surface illumination and back-scattered light is analyzed by a signal processor. According to described procedures, thinner sections of cartilage is found by studying intensity quotas for selected wavelengths, and arthritic/diseased cartilage is seen by studying changes in polarisation of the initial light. Two approaches are presented, one describing a single point measurement and one describing an imaging technique.

Description of the Drawings

Fig. 1 is the Monte Carlo model used for simulations including a cartilage layer of known thickness on top of a semi-infinite bone layer containing blood. The number of photons (out of incident 10^6) back-scattered to a ring shaped detector is calculated.

Fig. 2 upper graph shows mean reflection spectra from the reference material, including cartilage (dark line, n = 23) and bone containing blood (bright line, n = 10). Whiskers show ± SD. Lower graph shows reference spectrum of blood (TAKATANI and GRAHAM, 1979).

Fig. 3 shows examples of reflectance spectra from a grinding session. Before grinding (d = 1.66 mm), after first grinding (d = 0.78 mm), after third grinding (d = 0.44 mm) and after fourth grinding (d = 0 mm). Reference spectra of cartilage and bone containing blood are included for comparison. The spectra are separated in the y-direction for clarity.

Fig. 4 shows spectroscopically estimated cartilage thickness measure (d_{spec}) plotted against reference cartilage thickness for the complete material.

Fig. 5 shows Monte Carlo simulation results, including detected photons out of 10^6 ejected (in per cent) for increasing cartilage layer thicknesses. The simulations are performed for the single wavelength 633 nm.

Fig. 6 is a schematic drawing of a suggested arthroscopic probe and

Fig. 7 is a schematic drawing of the suggested arrangement.

Detailed description

Materials and methods

Twelve hip joint condyles from bovine calves were obtained from a local slaughterhouse less than 24 hours after sacrifice. Two of the condyles were used for reference measurements and the other ten for thickness experiments. The condyles were stored in saline in a refrigerator and prepared for cartilage measurements through the removal of soft tissues and tendons surrounding the joint.

Three sites on each condyle surface were used for the measurements. A handheld, rotating, grinding machine was used to reduce the cartilage layer thickness. Sandpaper with the roughness P100 was used for grinding. Care was taken to grind in short episodes (5 – 15 s) so as not to increase the temperature of the cartilage. Thickness measurement of the cartilage layer was done with a high-resolution ultrasound scanner (B-mode 20MHz, Dermascan 3v3, Cortex Technology, Hadsund, Denmark). The probe scanned over the measurement site and an image of the cartilage/bone interface was presented on the computer screen.

Optical reflection spectra were recorded by using an Oriel Instaspec IV CCD spectrometer equipped with an Ocean Optics broad spectrum tungsten lamp HL 2000 (spectral range 360 – 2000 nm). The light was guided by optical glass fibre bundles (NA = 0.35) and the measurements were taken at a small distance (2 – 5mm) to the condyle surface. The bundles were arranged in a probe head (diameter 4 mm) with the emitting fibre bundle encircling the detecting bundle. The reflection spectra were calculated according to the formula:

$$I(\lambda) = \frac{I_{tissue} - I_{background}}{I_{reference} - I_{background}} \qquad (1)$$

where I_{tissue} is the raw spectrum of the examined tissue, $I_{background}$ the detector background signal, and $I_{reference}$ the diffuse reflectance spectrum taken from a white reference ($BaSO_4$).

For each measurement position, spectra and ultrasound images were recorded from the intact cartilage layer, for 4 - 5 intermediate cartilage thicknesses (obtained by grinding) and when bone level had been reached.

Twenty-three pieces of pure cartilage, about 1 mm thick, were removed from the two joints by using a sharp knife. Remains of subchondral bone were carefully removed to secure a pure cartilage sample. Reflection spectra were measured, with the equipment described above, for each piece of cartilage placed on a black plastic sheet. The joints were cut in half, washed in saline and stored (in saline) for a few days to remove remains of blood. Finally, reflection spectra were taken from 10 positions on the exposed bone samples. Mean reference spectra for cartilage (n = 23) and bone (n = 10) were calculated and will be referred to as $S_{cartilage}$ and S_{bone}, respectively. Furthermore, the reference spectrum of blood (S_{blood}) was estimated as the inverse absorption spectrum of oxyhaemoglobin, taken from the literature (TAKATANI and GRAHAM, 1979). To decrease the influence of remains of blood in the bone, S_{bone} was adjusted by subtracting S_{blood} until the characteristic haemoglobin peaks could no longer be distinguished.

Each measured reflectance spectrum was matched to the true cartilage thickness (d), as determined from the stored ultrasound images. For the thickness determination, the manufacturer's software including a cursor system was used. The resolution of the ultrasound image was 0.06 mm. As a measure of cartilage thickness, determined from the spectroscopic data, d_{spec} was defined as the percentage contribution (%) of cartilage spectrum in the measured reflectance spectrum:

$$d_{spec} = \frac{a}{a+b+c} \tag{2}$$

where a, b and c are the coefficients for optimal match (least square fitting) between the measured spectrum ($S_{measured}$) and the reference spectra according to:

$$S_{measured} = a \cdot S_{cartilage} + b \cdot S_{bone} + c \cdot S_{blood} \tag{3}$$

An exponential regression model (4) was used for statistical comparison between reference cartilage thickness and d_{spec}.

$$d_{spec} = K_1(1 - e^{-K_2 \cdot d}) \tag{4}$$

where K_1 and K_2 are constants.

In order to test the assumption of an exponential relation between the cartilage thickness d and the corresponding spectroscopic data, a Monte Carlo model was used (Fig. 1). The model has a cartilage layer of known thickness (d) and diffusion theory optical properties (μ_{ac}, μ_{sc}, g_c) positioned on top of a semi-infinite layer of bone containing blood (μ_{ab}, μ_{sb}, g_b). The optical properties of cartilage, bone and blood were taken from the literature (BEEK et al., 1997; FIRBANK et al., 1993; TUCHIN, 2000) for the single wavelength 633 nm (see discussion). The optical properties of bone containing blood were set to the bone coefficients increased by those of blood at a selected perfusion level (10%) according to:

$$\mu_{ab} = \mu_{a,bone} + 0.10 \cdot \mu_{a,blood} \tag{5.1}$$

$$\mu_{sb} = \mu_{s,bone} + 0.10 \cdot \mu_{s,blood} \tag{5.2}$$

$$g_b = g_{bone} + 0.10 \cdot (g_{blood} - g_{bone}) \tag{5.3}$$

All optical parameters are presented in Table 1. Refractive indices of both layers were set to 1 as specular effects were not of interest. The pathways of 10^6 photons, incident in a point at the cartilage surface, were calculated (DE MUL et al., 1995). The back-scattered photons reaching a ring shaped detector (outer radius 3 mm, inner radius 1 mm) were counted. Simulations were performed for cartilage thicknesses d = 0 – 3 mm in steps of 0.1 mm.

Table 1 Tissue optical properties used in the Monte Carlo simulations

	Absorption coefficient μ_a [mm^{-1}]	Scattering coefficient μ_s [mm^{-1}]	Anisotropy factor g
Cartilage	0.033	21.4	0.909
Bone	0.040	35.0	0.925
Blood	1.60	413	0.997
Bone containing blood	0.20	76.3	0.932

Results

The mean reflection spectra of the reference material are presented in Fig. 2. The cartilage spectrum appears relatively "white" with a shift towards the blue region, whilst the bone spectrum appears distinctly "red" and includes the characteristic absorption peaks of haemoglobin at 542 nm and 576 nm. In the thickness analysis, the latter spectrum is divided into S_{bone} and S_{blood} (see Equation 3).

The mean (± SD) thickness of intact cartilage was 1.21 ± 0.30 mm (n = 30). A typical example of spectra from a grinding sequence is shown in Fig. 3. As the cartilage layer gets thinner a clear influence of bone can be seen. The spectroscopic estimation of cartilage thickness (d_{spec}) is plotted against the ultrasound reference cartilage thickness in Fig. 4. The regression model of Equation 4 is used ($r = 0.69$, $p < 0.000001$, $s = 0.167$, $K_1 = 0.75$, $K_2 = 3.81$, $n = 182$). For thinner cartilage layers ($d < 0.5$ mm), the model mean error is 0.19 ± 0.17.

The Monte Carlo simulation results are presented in Fig. 5. This result is similar to the experimental results in Fig. 4, with more photons reaching the detector at a thicker cartilage layer. The Monte Carlo simulation results support the assumption of an exponential relationship between cartilage thickness and the spectroscopic data.

Discussion

The main finding of this study is the possibility to extract objective information about cartilage thickness by studying the reflectance spectrum from the cartilage surface. The result implicates that it is possible to use a minimally invasive technique to characterise cartilage in connection with *in situ* diagnosis.

The reflectance spectrum from a condyle surface can be seen as a sum of spectra from cartilage and subchondral bone (containing blood). Typical spectra from cartilage and bone can be seen in Fig. 2. Cartilage contains relatively few cells which occupy 10-20% of its volume. The remainder is extracellular material which is highly hydrated and contains up to 80% water by weight. The material consists primarily of large hydrated proteoglycan aggregates, entrapped within a matrix of collagen fibrils. This fibre structure and the fact that absorption of water in the investigated wavelength region is low, gives reason to believe that the character of the cartilage spectrum is an effect of reduced scattering at longer wavelengths according to the Mie theory. Consequently, the cartilage layer acts as a diffuse reflector for photons, impeding them from reaching the highly absorbing subchondral bone.

We intended to assess cartilage thickness by studying the relative content of cartilage and bone components in the combined reflectance spectrum. By using this approach, the accuracy of cartilage thickness determination only becomes dependent upon the variability of the optical properties of these components. This variability remains to be investigated for a larger human material, after which it also will be possible to model the behavior for the complete spectrum and not just for a single wavelength. The exponential relationship between cartilage thickness and diffuse reflectance was expected (NÖTZLI et al., 1989).

A tentative source of error could be the variability of perfusion in the underlying bone. Flow rates in (rabbit) tibial and femoral cortical bone vary in a physiological range of 1.6 – 7.0 $ml \cdot min^{-1} \cdot g^{-1}$ (SHEPHERD and ÖBERG, 1990). Increased perfusion/blood content probably leads to increased absorption by the haemoglobin of the bone. Some of the data scattering in Fig. 4 may be due to the resolution of the ultrasound reference

system (0.06 mm according to the manufacturer's specifications). This value can be compared to the model mean error for thinner cartilage layers 0.19 ± 0.17 mm.

The accuracy of the ultrasound reference method depends on the ultrasound speed in cartilage (JURVELIN et al., 1995). Without a calibration of the device to cartilage ultrasound speed data we may have some spreading in data. Grinding causes roughening of the cartilage surface, affecting the degree of specular reflection. However, the specular reflection can be considered wavelength independent, not affecting the thickness estimation, solely based on specular distribution. For the same reason, the measurement distance was not precisely controlled.

To achieve maximal penetration depth, we chose to record the spectrum between 330 – 835 nm. It is reasonable to assume that a few characteristic wavelengths can be found that can be used for thickness calculations, thereby eliminating the need for recording the complete spectrum. Such an approach can facilitate the design of a future instrument based on the principle presented in the paper. An attractive feature of this principle is that it is based on fibre optics. Thus, it can probably be "imbedded" in an arthroscope, which, in addition to the visual assessment of the cartilage surface can give quantitative information about the thickness of the cartilage layer under study. Future challenges can involve the assessment of various aspects of cartilage quality such as the degree of fibrillation and surface roughness through fibre optic sensing.

We found a large variation in the thickness of the bovine hip joint cartilage (0.67 - 1.98 mm). The same variation can be found in human hip joint cartilage (1.14 - 2.84 mm), depending on where on the joint the cartilage is measured (NAKANISHI et al., 2001). There are reasons to believe that hip joints can be assessed through the sterile introduction of a fibre optic bundle but the most interesting application for this new principle may be the assessment of the knee via arthroscopy. The cartilage thickness of healthy and osteoarthritic human knee joints varies in the range of 0.5 to 7.4 mm (KLADNY et al., 1999). With the present method the haemoglobin absorption peaks could often be seen for thicker cartilage layers (Fig. 3) but a clear spectral effect occurred at cartilage thicknesses below 0.5 mm. In a well-perfused bone the variation and sensitivity of the method may be improved. Penetration depth may also be improved by focusing on the diffuse reflection component, by using more efficient optical components and by geometrical separation of light source and detector. Blood perfusion of bone has, for instance, been measured at 3.5 mm penetration depth, including

penetration of a 1 mm thick cartilage layer, using a 632.8 nm laser Doppler technique (NÖTZLI et al., 1999).

Conclusions

After studying bovine hip joint condyle surfaces, it was found that information about cartilage thickness could be extracted using optical reflectance spectroscopy. For thicker cartilage layers, a high reflection for the wavelengths 400-600 nm was seen, and for thinner cartilage layers, the characteristic spectra of blood and bone dominated. Consequently, the optical reflectance spectrum may be used to characterise cartilage, and specifically cartilage thickness, in connection with *in situ* diagnosis.

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Technical solution 1

Single point measurement by white light illumination, spectroscopic detection and signal processing for calculating intensity ratios to determine cartilage thickness. Suitable ratios are λ_1/λ_{ref} or λ_2/λ_{ref}, where λ_{ref} is a reference wavelength, possibly 630 nm, λ_1 is a haemoglobin absorption peak (425, 542 or 576 nm) and λ_2 is an infrared wavelength with high water absorption, possibly 1100 nm.

The solution is presented in Fig. 6 and Fig. 7. White light is supplied by a light source (15) emitting light into fiber bundle 1 (3). The light source is driven by a light source driver unit (14), stabilized by a control unit (13). The light source could be a broad band tungsten lamp. All fiber bundles could consist of high aperture optical glass fibers. Fiber bundle 1 (3) is passing through a channel (5) in the extension (9) of the arthroscopic probe (10) and supplies illumination of the measurement object (11) via a lens (6). This light serves both as illumination for investigation and for measurement. Light reflected from the object (11) is collected via the lens (6) into fiber bundle 2 (8). Fiber bundle 2 (8) passes through the same channel (5) and the intensity of light is measured by a detection unit (16) after manual input via an input device (12) requesting measurement to start. The detection unit is located in a control apparatus (19) and could be a CCD spectrometer. The detected signal is processed in a signal processor (17) according to the theory presented above. Cartilage thickness result is presented on a display unit (18). One solution of the arthroscopic probe (10) includes channels for saline perfusion and suction (4) and an ocular channel (1) for visual observation through an eyepiece (7) during the measurement. The visual image can be focused using a screw (2).

Technical solution 2

Single point measurement by discrete wavelength illumination, detection and signal processing for calculating intensity ratios to determine cartilage thickness. Chosen wavelengths could be λ_1, λ_2 and λ_{ref}, giving ratios λ_1/λ_{ref} and λ_2/λ_{ref}, where λ_{ref} is a reference wavelength, possibly 630 nm, λ_1 is a haemoglobin absorption peak (425, 542 or 576 nm) and λ_2 is an infrared wavelength with high water absorption, possibly 1100 nm.

The solution is presented in Fig. 6 and Fig. 7. Both white light and discrete wavelength light are supplied by light sources (15) emitting light into fiber bundle 1 (3). The light sources are driven by a light source driver unit (14), stabilized by a control unit (13). The light sources could be a broad band tungsten lamp and stable light emitting diodes. All fiber bundles could consist of high aperture optical glass fibers. Fiber bundle 1 (3) is passing through a channel (5) in the extension (9) of the arthroscopic probe (10) and supplies illumination of the measurement object (11) via a lens (6). The white light serves as illumination for investigation and the discrete wavelength light for measurement. Light reflected from the object (11) is collected via the lens (6) into fiber bundle 2 (8). Fiber bundle 2 (8) passes through the same channel (5) and the intensity of light is measured by a detection unit (16) after manual input via an input device (12) requesting measurement to start. At measurement start, the white light source is turned off, as controlled by the control unit (13). The detection unit (16) is located in a control apparatus (19) and could consist of photo diodes. The detected signal is processed in a signal processor (17) according to the theory presented above. Cartilage thickness result is presented on a display unit (18). One solution of the arthroscopic probe (10) includes channels for saline perfusion and suction (4) and an ocular channel (1) for visual observation through an eyepiece (7) during the measurement. The visual image can be focused using a screw (2).

Technical solution 3

Imaging by using white light illumination, optical filters and/or signal processing for creating images with enhanced contrast between cartilage and bone, and/or contrast between fibrillated and healthy cartilage. Chosen wavelengths could be λ_1, λ_2 and λ_{ref} giving ratios λ_1/λ_{ref} or λ_2/λ_{ref}, where λ_{ref} is a reference wavelength, possibly 630 nm, λ_1 is a haemoglobin absorption peak (425, 542 or 576 nm) and λ_2 is an infrared wavelength with high water absorption, possibly 1100 nm.

The solution is presented in Fig. 6 and Fig. 7. White light is supplied by a light source (15) emitting light into fiber bundle 1 (3). The light source is driven by a light source driver unit (14), stabilized by a control unit (13). The light source could be a broad band tungsten lamp. The fiber bundles could consist of high aperture optical glass fibers. Fiber bundle 1 (3) is passing through a channel (5) in the extension (9) of the

arthroscopic probe (10) and supplies illumination of the measurement object (11) via a lens (6). This light serves both as illumination for investigation and measurement. Light reflected from the object (11) is collected via the lens (6) into fiber bundle 2 (8). In one solution polarization filters are included at the fiber tip for the measurement of cartilage fibrillation. Fiber bundle 2 (8) passes through the same channel (5) and the intensity of light is measured by a detection unit (16) after manual input via an input device (12) requesting measurement to start. The detection unit (16) is located in a control apparatus (19) and could consist of a 2D CCD array. In one solution, the detection device includes optical filters before detection, and in another solution, image processing is performed digitally after detection by a signal processor (17), in both cases according to the theory presented above. Contrast enhanced images of the measurement object (11) is presented on a display unit (18). One solution of the arthroscopic probe (10) includes channels for saline perfusion and suction (4) and an ocular channel (1) for visual observation through an eyepiece (7) during the measurement. The visual image can be focused using a screw (2).

CLAIMS

1. Device for measuring joint cartilage qualities, such as cartilage thickness, cartilage surface roughness and degree of cartilage fibrillation, comprising an arthroscopic probe (19) with an extension (9) for inspection of the joint, wherein fiber bundles (3, 8) are arranged in said extension c h a r a c t e r i z e d in

that the fibre bundles include a first set of fibers (3) for conveying light from a light source (15) to illuminate the joint surface and a second set of fibers (8) for conveying light reflected from the joint surface to a detector means (16),

that said detector means (16) is designed for measuring the intensity of light reflected from the joint surface.

2. Device in accordance with claim 1, wherein said detector means (16) is a light intensity detector for single wavelengths or for a spectrum of wavelengths, that is connected to a signal processor (17), provided in a control apparatus (19), said signal processor being configured to apply a cartilage thickness algorithm on data acquired from said detector means (16).

3. Device in accordance with claim 2, wherein said cartilage thickness algorithm utilizes the fact that the photon absorption of cartilage and subchondral bone is different at wavelength regions related to blood chromophores or water.

4. Device in accordance with claim 1, wherein said detector means (16) is a light intensity detector for single wavelengths or for a spectrum of wavelengths, sensitive to polarization state of measured light, that is connected to a signal processor (17), provided in a control apparatus (19), said signal processor being configured to apply a cartilage fibrillation algorithm on data acquired from said detector means (16).

5. Device in accordance with claim 4, wherein said cartilage fibrillation algorithm utilizes the fact that the polarization states of photons are different after back-scattering by healthy or diseased/arthritic cartilage.

6. Device for imaging joint surfaces, enhancing contrast between healthy and diseased/arthritic regions, including thin cartilage regions, regions with rough surface cartilage and regions with highly fibrillated cartilage, comprising an arthroscopic probe (19) with an extension (9) for inspection of the joint, wherein fiber bundles (3, 8) are arranged in said extension c h a r a c t e r i z e d in

that the fibre bundles include a first set of fibers (3) for conveying light from a light source (15) to illuminate the joint surface and a second set of fibers (8) for conveying light reflected from the joint surface to an two-dimensional detector means (16),

that said two-dimensional detector means (16) is designed to present the intensity of light reflected from the joint surface.

7. Device in accordance with claim 6, wherein said two-dimensional detector means (16) is a light intensity detector for single wavelengths or for a spectrum of wavelengths, that is connected to a signal processor (17), provided in a control apparatus (19), said signal processor being configured to apply a cartilage thickness algorithm on data acquired from said two-dimensional detector means (16).

8. Device in accordance with claim 7, wherein said cartilage thickness algorithm utilizes the fact that the photon absorption of cartilage and subchondral bone is different at wavelength regions related to blood chromophores or water.

9. Device in accordance with claim 6, wherein said two-dimensional detector means (16) is a light intensity detector for single wavelengths or for a spectrum of wavelengths, sensitive to polarization state of measured light, that is connected to a signal processor (17), provided in a control apparatus (19), said signal processor being configured to apply a cartilage fibrillation algorithm on data acquired from said two-dimensional detector means (16).

10. Device in accordance with claim 9, wherein said cartilage fibrillation algorithm utilizes the fact that the polarization states of photons are different after back-scattering by healthy or diseased/arthritic cartilage.

11. Method for measuring joint cartilage qualities, such as cartilage thickness, cartilage surface roughness and degree of cartilage fibrillation, according to claims 1-10.

ABSTRACT

The invention described is an arrangement and a method for measuring joint cartilage qualities, such as cartilage thickness, cartilage surface roughness and degree of cartilage fibrillation. The arrangement comprises an arthroscopic probe (19), fiber bundles (3, 8), a light source (15) for emitting light, a detection unit (16) for detecting reflected light, and a signal processor (17) for processing the detected signals. The arrangement utilizes a new principle for cartilage layer thickness assessment in joints, based on the differences in absorption spectrum between cartilage and subchondral bone, and a new principle for cartilage fibrillation assessment, based on the differences in photon polarization state between light back-scattered from healthy or diseased/arthritic cartilage.

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Figure 1

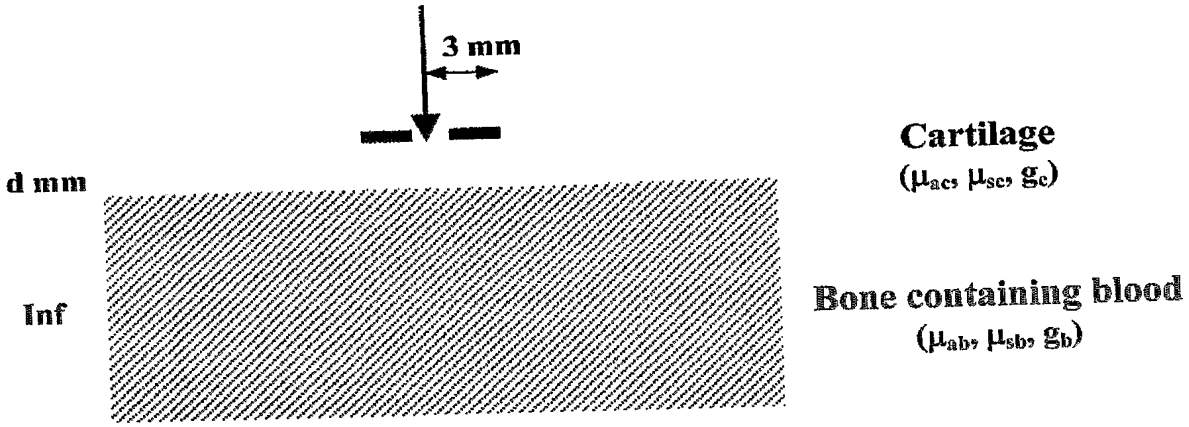


Figure 2

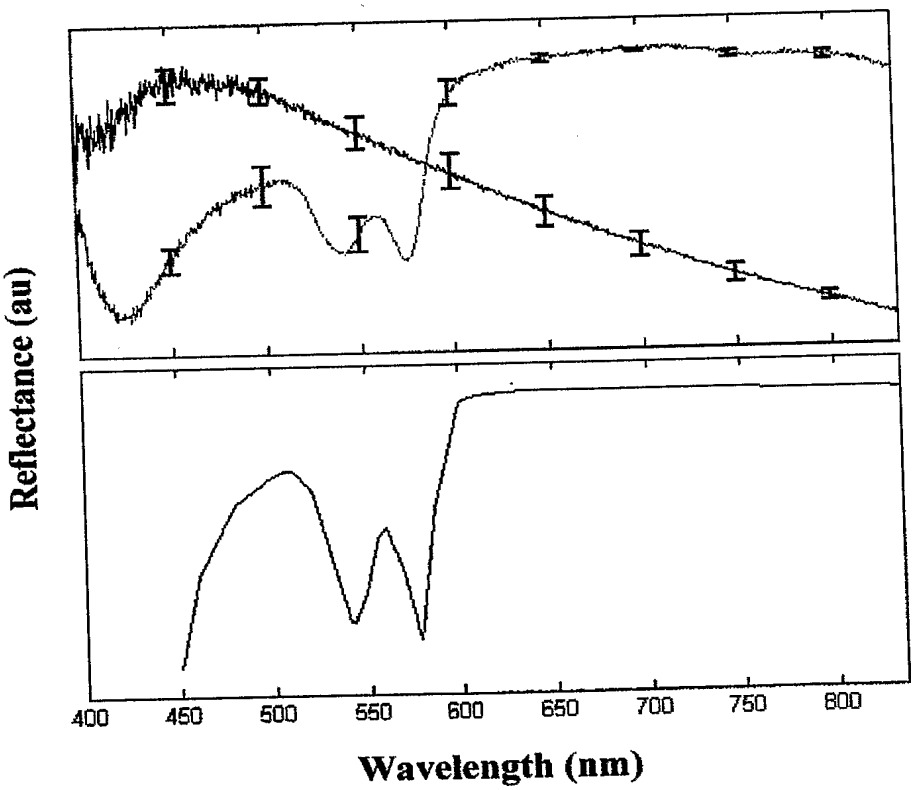
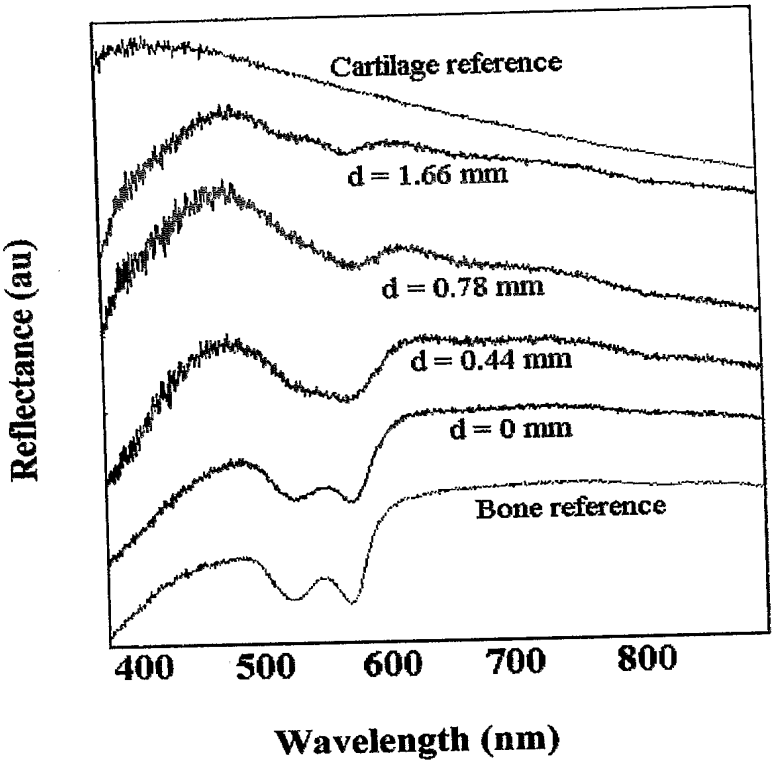


Figure 3



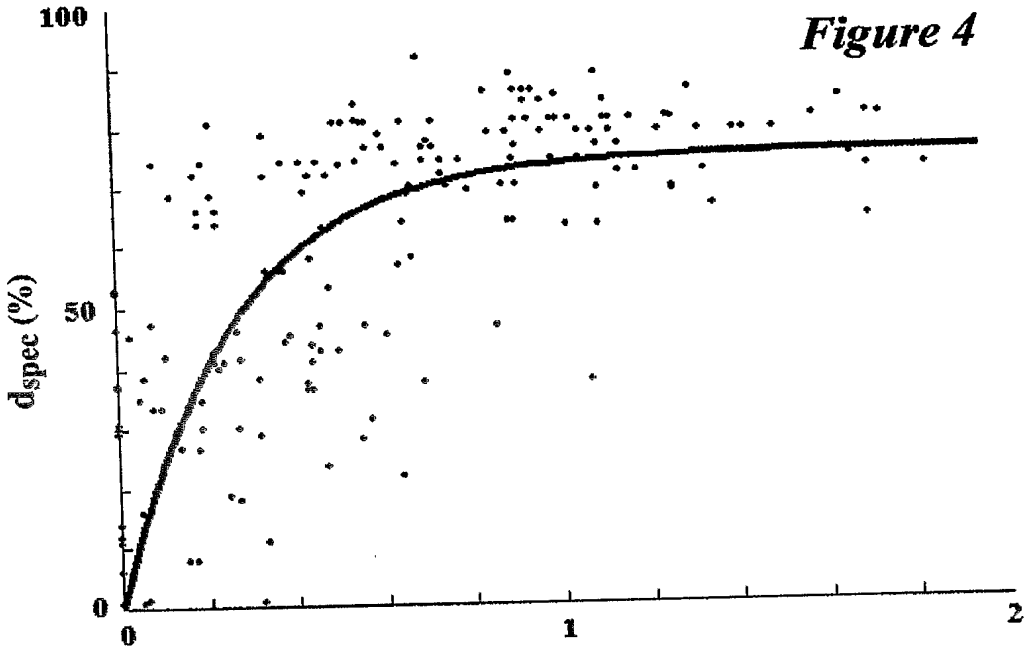


Figure 4

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Figure 5

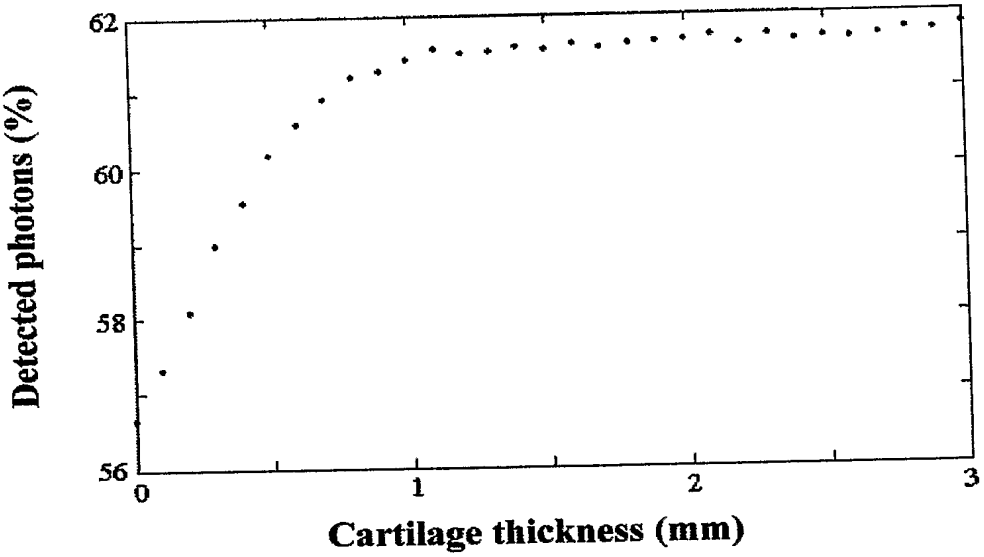


Figure 6

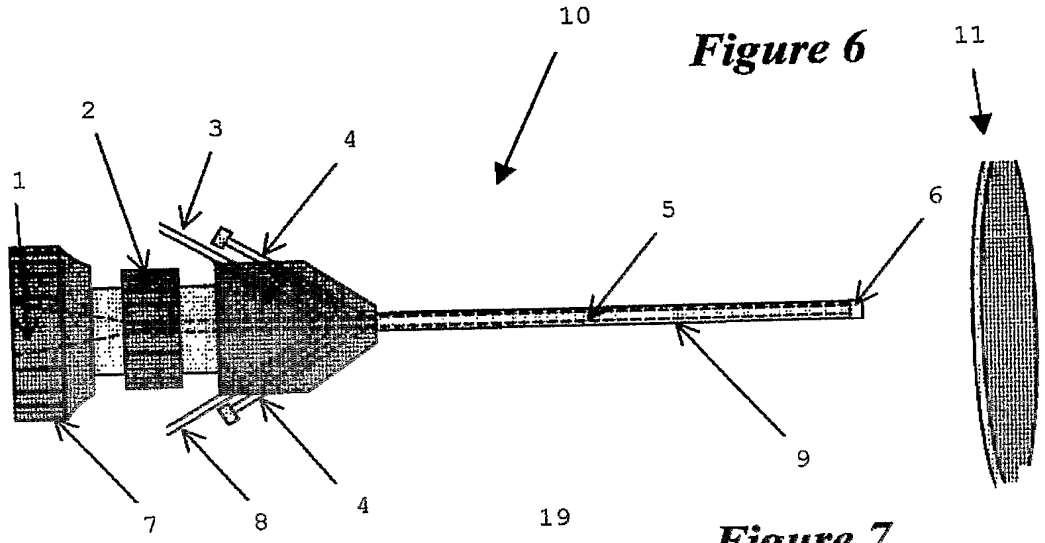
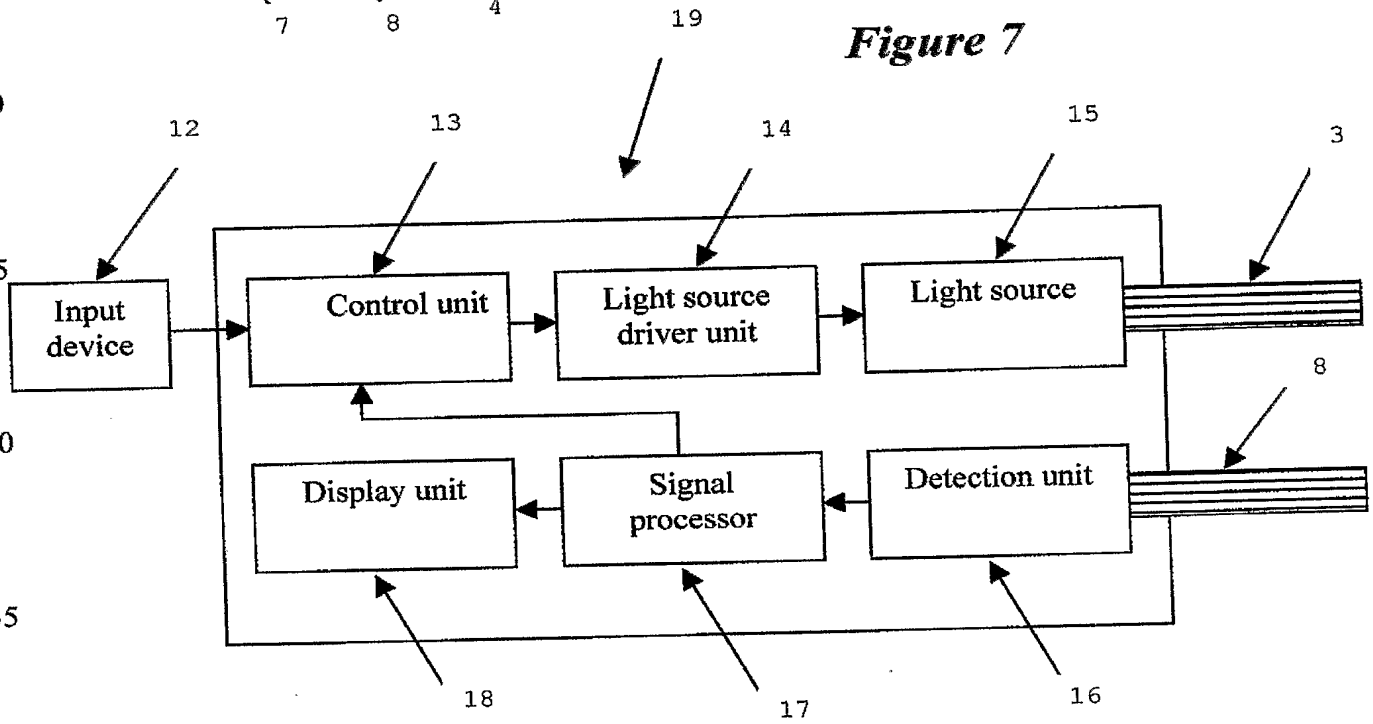


Figure 7



1	Ocular channel	11	Measurement object
2	Focusing screw	12	Input device
3	Fiber bundle 1	13	Control unit
4	Channels for saline perfusion/suction	14	Light source driver unit
5	Channels through extension	15	Light source
6	Lens	16	Detection unit
7	Eyepiece	17	Signal processor
8	Fiber bundle 2	18	Display unit
9	Extension	19	Control apparatus
10	Arthroscopic probe		